Biomechanical Comparison of the Docking Technique With and Without Humeral Bioabsorbable Interference Screw Fixation

Jason G. Hurbanek,* MD, Kyle Anderson,†‡ MD,
Stephanie Crabtree,§ BS, and G. Joshua Karnes,§ MS
*From *Henry Ford Hospital, Detroit, Michigan, ‡William Beaumont Hospital, Royal Oak, Michigan, and §Arthrex, Inc, Naples, Florida*

Background: Surgical reconstruction of the ulnar collateral ligament has evolved since Frank Jobe's original description. The "docking technique" is a popular modification that allows for securing the graft within a single humeral tunnel. More recently, interference screw fixation has been introduced as a means of improving the ultimate strength, stiffness, and kinematics of these constructs.

Purpose: This study was conducted to compare the biomechanical performance of the docking technique with and without interference screw fixation in the humerus.

Study Design: Controlled laboratory study.

Methods: Nine matched pairs of human cadaveric elbows (age 49.9 ± 8.0 years) were reconstructed with a tendon graft using the docking technique (group 1) or the docking technique with the addition of a 4.75-mm bioabsorbable humeral interference screw (group 2). Before the reconstruction, joint laxity was measured on each specimen with the ulnar collateral ligament intact and then after transection of the ligament. Laxity measurements were repeated after the reconstruction. Failure testing was then performed at 70° of elbow flexion. The specimens were preloaded with a 1-N·m moment and then loaded to failure at a displacement rate of 14 mm/s to approximate 50% strain per second.

Results: Within group 1, the elbow laxity of the reconstructed state was significantly greater than the intact state at all tested flexion angles ($P < .021$). Within group 2, no statistically significant difference existed in elbow laxity between the intact state and the reconstructed state. When comparing laxities between groups, group 1 tended to be more lax at all tested flexion angles but was only significantly greater at 105° of flexion. The most common mode of failure for both groups involved the sutures pulling out of the tendon. No significant difference was found for ultimate moment of failure between the 2 groups. However, the moment associated with 3 mm of gap formation for group 2 (12.8 ± 4.2 N·m) was statistically greater than that of group 1 (7.5 ± 1.2 N·m) ($P = .001$). The stiffness of group 2 (14.7 ± 6.4 N/mm) was significantly greater than group 1 (9.9 ± 3.1 N/mm) ($P = .044$).

Conclusion: The biomechanical performance of the docking technique with and without a humeral interference screw is similar.

Clinical Relevance: The stiffness of the construct, along with the difference in moment that allows a 3-mm gap formation, suggests that the addition of a humeral interference screw is potentially beneficial. Further research in a healing model will help clarify this benefit.

Keywords: ulnar collateral ligament reconstruction; docking technique; interference screw; biomechanical

The ulnar collateral ligament (UCL) is an important structure for resisting valgus force at the elbow.[4,10,12,16,17] The UCL consists of 3 bundles: anterior, posterior, and transverse. The primary stabilizing structure of the elbow is the anterior bundle of the UCL, and it is functionally subdivided into 2 bands (anterior and posterior). The anterior bundle originates from the anteroinferior surface of the medial epicondyle of the humerus and inserts on the sublime tubercle of the ulna.[4]

The UCL is at risk for injury in overhead athletes due to large valgus forces generated during the late cocking and

†Address correspondence to Kyle Anderson, MD, 27207 Lahser Road #200B, Southfield, MI 48034 (e-mail: kanderson_md@hotmail.com).

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early acceleration phases of the throwing motion.[23] These forces have been estimated to range from 64 to 120 N·m and are resisted primarily by the anterior bundle of the UCL from 30° to 120° of elbow flexion.[4,8,23] In addition, the triceps, wrist flexors/pronators, and anconeus muscles contribute by acting as dynamic stabilizers.[23]

Ulnar collateral ligament reconstruction has evolved considerably since Dr Frank Jobe first described his technique.[12] This procedure, commonly known as "Tommy John surgery," involved exposing the UCL by detaching the flexor-pronator mass, transposing the ulnar nerve submuscularly, and passing an autologous tendon graft through bone tunnels on the medial epicondyle of the humerus and the ulnar sublime tubercle. The graft was then fixed in a figure-of-8 fashion and sutured to itself under tension.[12] Multiple authors have reported improved results with modifications to Jobe's first description.[3,5,22]

There have been several more recent modifications of the operation in an effort to reduce morbidity and improve rates of return to preinjury competition. Smith et al[21] defined a "safe zone" where the flexor-pronator mass could be split to avoid detaching it from the epicondyle. Rohrbough et al[20] reported on their series of patients who underwent reconstruction using the "docking technique." This technique involves a single humeral tunnel with suture fixation over a bony bridge on the superior aspect of the medial epicondyle of the humerus. They used the muscle- splitting approach, allowing them to avoid an obligatory transposition of the ulnar nerve. Other authors have experimented with the number of strands in the UCL graft within the docking construct. Koh et al[13] reported on a combination of 2- and 3-strand constructs within the docking technique, noting that 18 of 19 athletes returned to their previous level of participation at 13.1 months. Paletta and Wright[19] further modified the docking procedure by using a 4-strand palmaris longus graft; they reported a 92% return to preinjury level of competition at a mean of 11.5 months.

Multiple biomechanical studies have been performed to support the clinical use of these constructs, and the procedure continues to evolve. With the goal of improving the strength and biomechanics of the reconstruction relative to the native ligament while decreasing surgical invasiveness, Ahmad et al[1] examined the use of interference screw fixation in UCL reconstruction. They evaluated elbow kinematics in 10 matched pairs of cadaveric elbows. The UCL was reconstructed with autologous palmaris longus graft fixed with 5 × 15-mm soft tissue interference screws in single pilot holes on the medial epicondyle and sublime tubercle. The authors reported that the failure strength of the reconstructed elbow was comparable with that of the native ligament and that physiologic elbow kinematics were reliably restored.[1]

Other investigators have published results on biomechanical studies examining the use of interference screws in UCL reconstruction. Large et al[14] performed a biomechanical study comparing the Jobe technique, metal interference screws in both the ulna and humerus, and the native UCL. They concluded that the failure strength and initial and overall stiffness of the traditional Jobe bone-tunnel UCL reconstruction was superior to the metal interference screw construct and that the Jobe bone-tunnel technique reproduced the initial and overall stiffness of an intact UCL. McAdams et al[15] performed a cyclic loading biomechanical evaluation of elbows fixed with either the docking or a bioabsorbable interference screw procedure (on both the ulna and humerus). They found that the interference screw fixation resulted in less valgus angle widening in response to early cyclic valgus load when compared with the docking technique. Lastly, Armstrong et al[2] biomechanically compared 4 reconstruction techniques and showed that the docking and EndoButton (Smith & Nephew Endoscopy, Andover, Massachusetts) constructs, although significantly weaker than native ligaments, presented peak loads to failure significantly greater than those of the Jobe and interference screw methods.

The advantages of an interference screw construct over the Jobe or docking construct are not clear. Also, whether there is a more appropriate role for interference screws in ulnar fixation, humeral fixation, or both is unknown. The purpose of this study was to compare the biomechanical performance of UCL repairs using a docking technique with and without interference screws used specifically on the humerus alone. The hypothesis is that the docking technique with interference screws more closely replicates the kinematics of an intact UCL, and has higher stiffness than a docking construct alone.

MATERIALS AND METHODS

Nine matched pairs of human cadaver elbows (age 49.9 ± 8.0 years) were dissected of all soft tissue except the elbow capsule and ligaments. The humerus was cut 14 cm superior to the joint line and potted in 2-in schedule-40 polyvinyl chloride (PVC) pipe using fiberglass epoxy resin. The long axis of the humerus was aligned with the long axis of the PVC pipe. The radius and ulna were cut 14 cm inferior of the joint line and potted in 2-in PVC pipe in a neutral position. The length of the PVC pipe and the amount of resin used to pot the radius and ulna were kept constant for all specimens.

The palmaris longus tendon was harvested and used for reconstruction of the UCL. For specimens that did not have the palmaris longus, the flexor digitorum superficialis (FDS) was harvested and used (the diameter of each palmaris tendon was measured with a caliper, and if an FDS tendon was used, it was trimmed to the average size of the palmaris tendons that had previously been used). Two pairs of FDS tendons were used. These specimens were analyzed as part of the complete data set and not as a subgroup.

Kinematic Testing

A custom fixture, shown in Figure 1, was created to fix the humerus, allowing 5 degrees of freedom for the distal end (only the flexion angle was constrained). The humerus was fixed so that the medial side of the elbow was facing superiorly and aligned in the fixture so that the center line of the epicondyles was perpendicular to a horizontal plane. A 6-in level placed on the anterior aspect of the epicondyles was used to confirm alignment of the humerus.

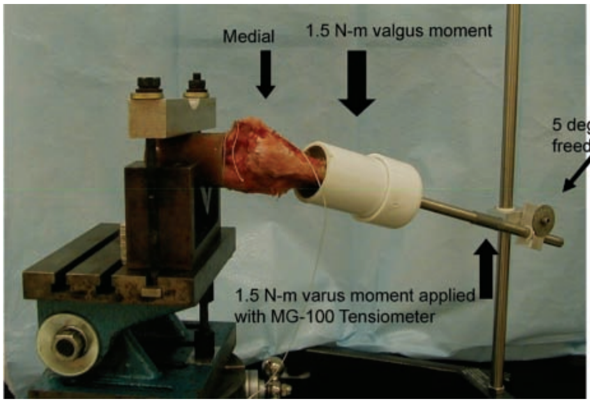

Figure 1. Kinematic test setup.

The weight of the fixture and potting material created a 1.5-N·m valgus moment across the elbow joint. Using the same test setup, a 1.5-N·m varus moment could be applied across the elbow joint by introducing a known upward-oriented load at the connection of the cylindrical rods that allow the 5 degrees of freedom. The load was applied using a MG-100 tensiometer (Mark-10 Corp, Copiague, New York), and the amount of load necessary was determined using free-body diagrams. The angle relative to a horizontal plane of the distal end of the construct was measured using a goniometer at each respective loading scenario to determine the elbow laxity at 0°, 15°, 30°, 45°, 60°, 75°, 90°, and 105° flexion angles.

The elbow laxity of the intact specimen was measured in degrees at all respective flexion angles. Before failure testing, the native length of the anterior bundle of the UCL was measured using calipers. After completion of the kinematic testing, the UCL was transected to simulate a defect, and the elbow laxity was measured at the 7 different flexion angles. The ligament was reconstructed, and elbow laxity was re-measured.

Ligament Reconstruction

Two versions of the docking procedure were used to reconstruct the UCL. Group 1 received a UCL reconstruction using the traditional docking procedure, and group 2 received a docking procedure with the addition of a 4.75-mm × 15-mm BioTenodesis screw (Arthrex, Naples, Florida) inserted in the humeral tunnel. In both instances, a custom guide was used to create 2 intersecting 4-mm tunnels in the ulna 5 mm distal to the joint line, as seen in Figure 2. This left a bony bridge approximately 8 to 10 mm wide. A 4-mm tunnel was created in the humerus, positioned in the central 2 quartiles or 50% of the medial epicondyle in the anterior position of the existing UCL as previously described.[19] It was then overdrilled to a diameter of 5 mm for a depth of 15 mm. A 1-mm drill bit was then used to create exit holes for the suture, bridged approximately 7 to 10 mm, at the proximal aspect of the humeral epicondyle. Approximately 1 cm of the free end of the tendon graft was whipstitched in a pattern resembling a multiple

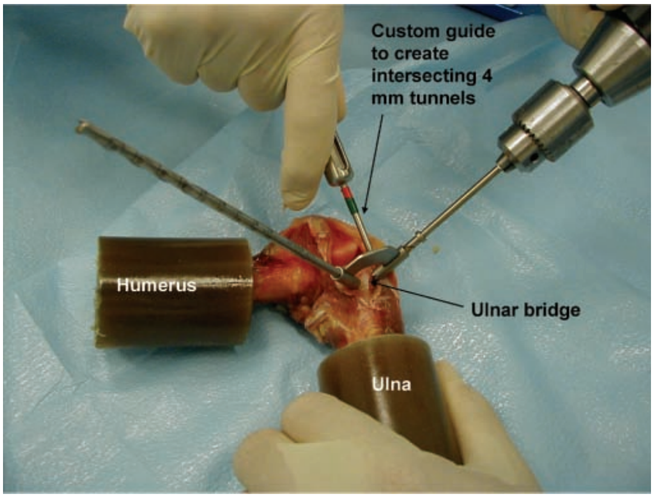

Figure 2. Ulnar tunnel preparation.

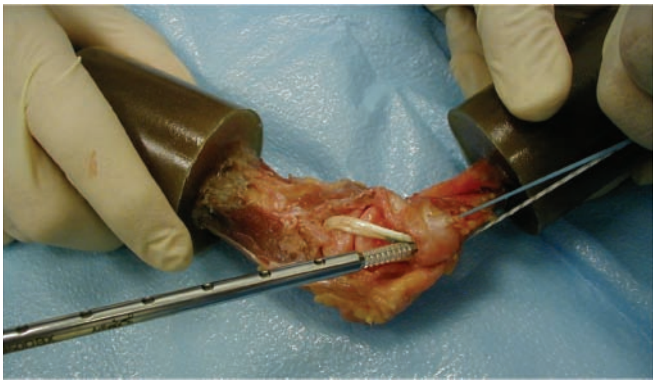

Figure 3. Group 2 specimen. Tension was placed on the FiberWire sutures, the elbow was held in 30° of flexion, and the BioTenodesis screw was inserted into the humeral tunnel. The sutures were then tied over the humeral epicondyle. Group 1 specimen grafts were docked and secured in the same fashion, but no screw was placed.

locking Krackow stitch, using free strands of No. 2 FiberWire (Arthrex). This suture technique was previously accepted.[1] The graft was passed through the ulnar tunnel and the posterior limb of the graft was pulled into the humeral tunnel. The remaining free end of the graft was prepared by estimating its final length by laying it adjacent to the humeral tunnel and marking the graft such that approximately 7 to 8 mm of graft would be free to enter the humeral tunnel. The graft was marked, and then whipstitched as described above. The remaining graft was amputated. The sutures were passed through the suture holes on the proximal portion of the epicondyle, securely docking both ends of the graft within the humeral tunnel. The graft was tensioned with the arm held at 30° of flexion with a manually applied varus moment at the elbow while holding the forearm in full supination. Group 2 differed from group 1 by placement of a 4.75 × 15-mm BioTenodesis screw in the humeral tunnel while applying tension to the graft via the sutures (see Figure 3). The sutures were then

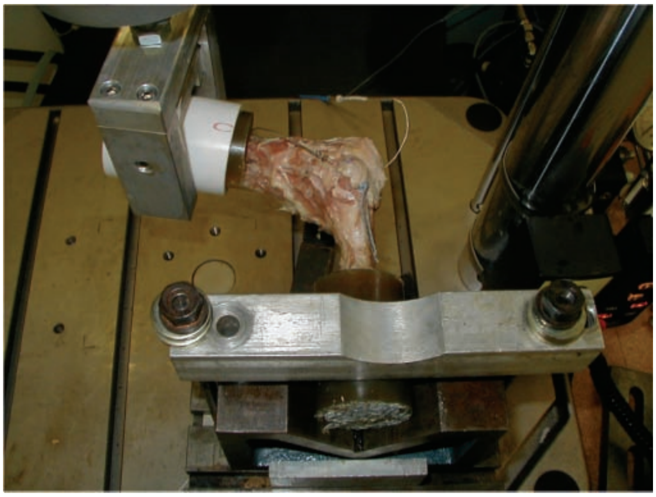

Figure 4. Failure testing setup.

tied over the bony bridge. The screw was placed before tying the knots to avoid potential tissue relaxation before screw placement.

Failure Testing

The proximal end of the elbow construct was fixed in a custom fixture so that the medial side was oriented vertically, and the center line of the epicondyles was perpendicular to a horizontal plane. The flexion angle of the distal end of the construct was constrained at 70°; however, all other degrees of freedom were unconstrained. The failure testing setup is shown in Figure 4. Before failure testing, the initial gap at the joint line of the humerus and ulna was recorded and a differential variable reluctance transducer (Microstrain, Burlington, Vermont) was placed at the joint line to measure the gap formation during failure testing. After failure testing, the moment was calculated at 3 mm of gap formation. This distance was chosen based on clinical examination of instability.[11]

The load was applied to the distal end of the construct at a distance of 10 cm from the joint line. The specimens were preloaded so that a 1-N·m moment was applied across the joint line. The specimens were loaded to failure at a displacement rate at 14 mm/s to approximate 50% strain per second.[1] The mode of construct failure was noted.

Data Analysis

A pairwise Student *t* test ($\alpha = .05$) was used to compare the intact and repaired elbow laxity at each individual flexion angle. A sequential Bonferroni correction using the Simes-Hochberg method was performed to account for alpha inflation due to the multiple comparisons. Comparisons were made within specimens (eg, specimens that received a docking repair with a screw were compared to their respective intact state). It was assumed that the defect state would have increased elbow laxity compared with all other states; therefore, the defect state was not included in any statistical comparisons so that the power of the statistical tests was not decreased.

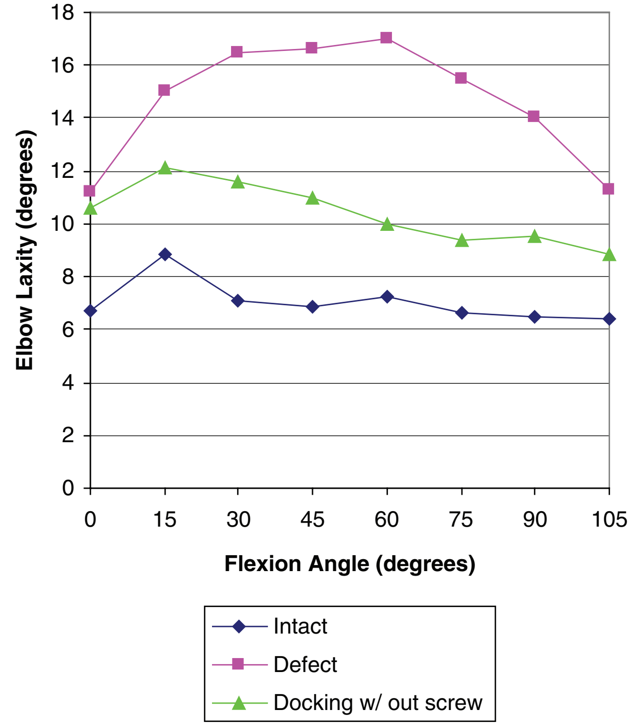

Figure 5. Elbow laxity—docking without screw significantly greater than intact at all flexion angles ($P < .021$).

A pairwise Student *t* test ($\alpha = .05$) was used to compare elbow laxity of the 2 reconstructions. The data were normalized by subtracting the elbow laxity of the intact state from the repaired state for each specimen. For example, the laxity of a specimen in the reconstructed state was subtracted from its intact laxity at a specific flexion angle. Differences for the remaining specimens were determined at that flexion angle. Those differences were averaged, generating a data point for analysis. A sequential Bonferroni correction using the Simes-Hochberg method was performed to account for alpha inflation due to the multiple comparisons.

The loads to failure of the 2 groups were compared using a pairwise Student *t* test ($\alpha = .05$). The construct stiffness, moment at failure, and moment at 3-mm gap formation were examined.

RESULTS

Within group 1 (specimens that received docking repair without a screw), the elbow laxity of the reconstructed state was significantly greater than that of the intact state at all flexion angles ($P < .021$ for all angles) (Figure 5 and Table 1). Within group 2 (docking repair with screw), no statistically significant difference existed in elbow laxity between the intact state and the reconstructed state (Figure 6 and Table 2). When comparing laxities between groups

TABLE 1

Elbow Laxity Data (in Degrees) for Group 1 (Docking Reconstruction Without Screw)[a]

Status	Flexion Angle							
	0°	15°	30°	45°	60°	75°	90°	105°
Intact	7 ± 4	9 ± 4	7 ± 4	7 ± 3	7 ± 2	7 ± 2	7 ± 2	6 ± 2
Defect	11 ± 5	15 ± 7	17 ± 7	17 ± 7	17 ± 8	16 ± 8	14 ± 6	11 ± 4
Reconstruction	11 ± 5	12 ± 4	12 ± 4	11 ± 4	10 ± 3	9 ± 3	10 ± 3	9 ± 2

[a]Each group 1 specimen (n = 9) underwent laxity testing in the intact, defect, and reconstructed state at all flexion angles. Data represent mean laxities and standard deviations for the tested specimens.

TABLE 2

Elbow Laxity Data (in Degrees) for Group 2 (Docking Reconstruction With Screw)[a]

Status	Flexion Angle							
	0°	15°	30°	45°	60°	75°	90°	105°
Intact	7 ± 5	9 ± 6	9 ± 6	9 ± 6	8 ± 5	7 ± 4	7 ± 4	7 ± 5
Defect	13 ± 5	15 ± 5	16 ± 8	16 ± 7	16 ± 7	16 ± 9	15 ± 9	12 ± 6
Reconstruction	9 ± 3	10 ± 3	10 ± 3	9 ± 3	8 ± 2	6 ± 2	6 ± 2	5 ± 3

[a]Each group 2 specimen (n = 9) underwent laxity testing in the intact, defect, and reconstructed state at all flexion angles. Data represent mean laxities and standard deviations for the tested specimens.

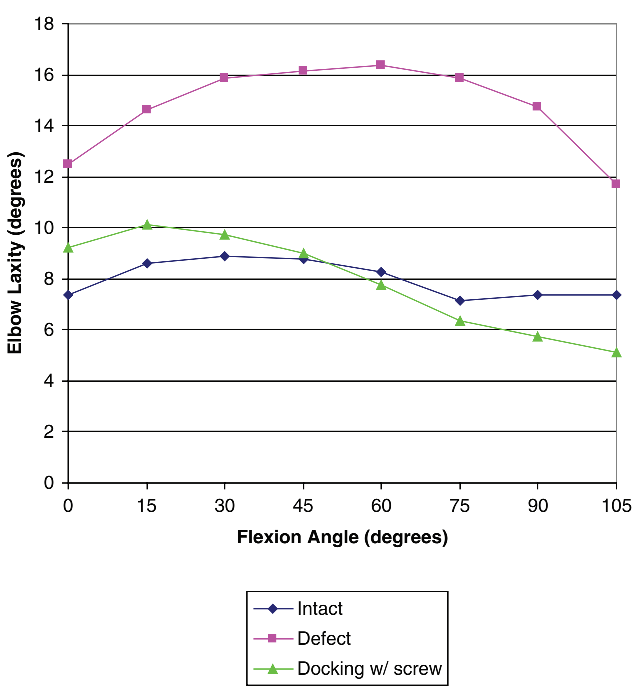

Figure 6. Laxity of docking with screw no different than intact.

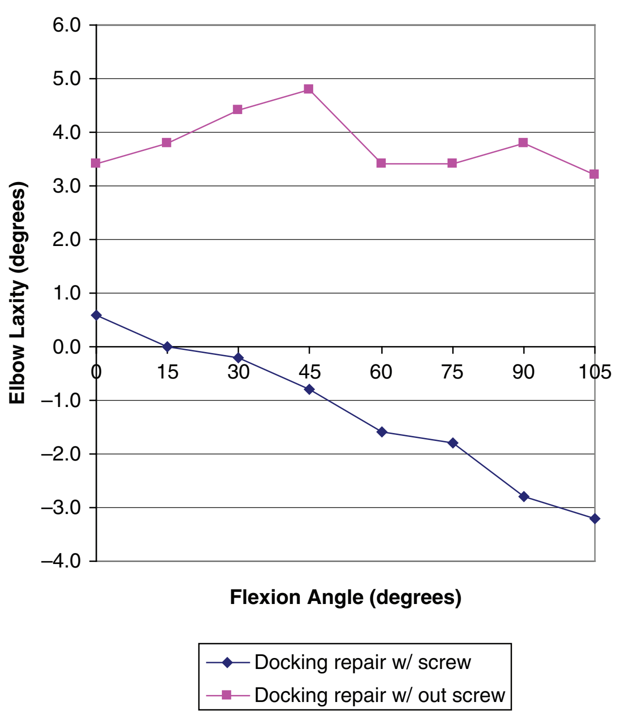

Figure 7. Elbow laxity normalized about intact status. Positive value indicates increased laxity after reconstruction, negative value indicates decreased laxity, and 0 indicates restoration of original state.

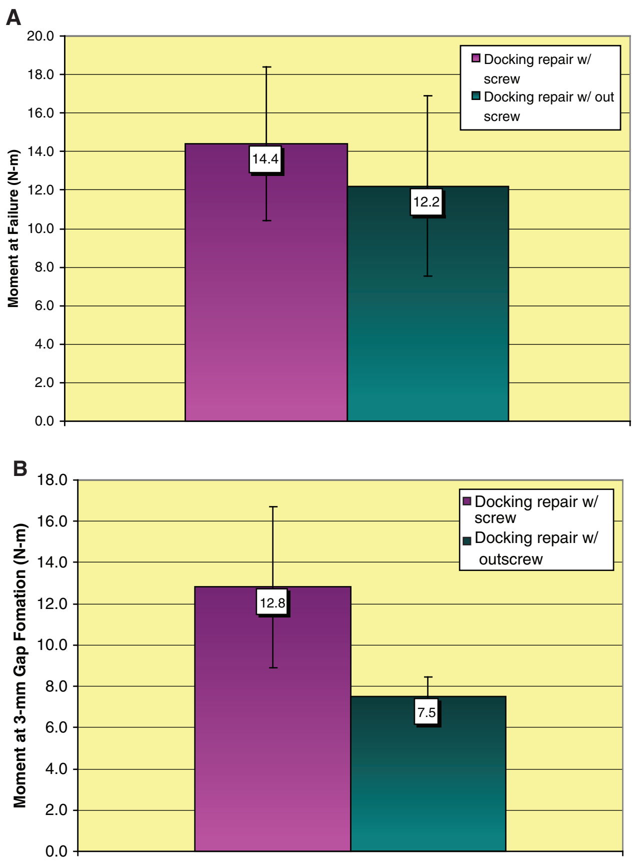

Figure 8. A, ultimate moment of failure. No significant difference existed (P = .161). B, moment at 3-mm gap formation. Significant difference detected between groups 1 and 2 (P = .001).

normalized about an intact status, group 1 demonstrated greater laxity at all tested flexion angles, and was only significantly greater than group 2 at 105° of flexion (P = .004) (Figure 7). Group 2 demonstrated a trend toward less laxity (more constrained) with increasing flexion angles.

The predominant mode of failure (n = 8) for group 1 was the whipstitches pulling through the tendon. The predominant mode of failure (n = 7) in group 2 was the whipstitches pulling through the tendon. Within these 7 failures, the tendon also slipped past the screw 4 times. One specimen in group 2 failed due to midsubstance rupture of the anterior tendon strand. One pair (from the same 57-year-old donor) in each group failed due to ulnar tunnel fracture.

The moment across the elbow joint at failure for group 2 was 14.4 ± 4.0 N·m. The moment at failure of group 1 was 12.2 ± 4.7 N·m. There was no statistical difference between the 2 groups (P = .161) (Figure 8A). The moment across the joint at 3 mm of gap formation for group 2 was 12.8 ± 3.9 N·m. The moment at 3 mm of gap formation for group 1 was 7.5 ± 1.0 N·m. This difference was significant (P = .001) (Figure 8B). The stiffness of group 2 was 14.7 ± 6.4 N/mm and for group 1 was 9.9 ± 3.1 N/mm. This difference was significant (P = .044) (Figure 9).

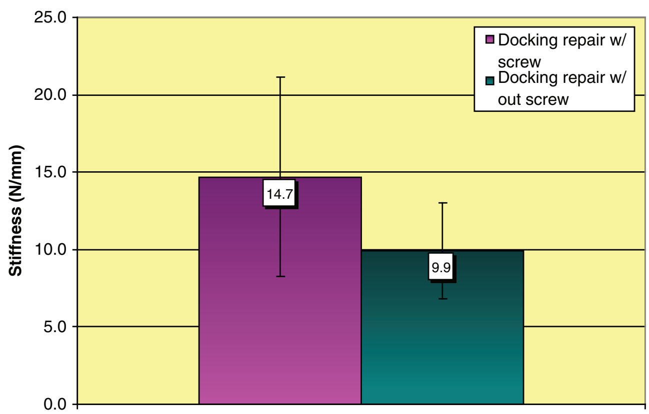

Figure 9. Stiffness. Groups 1 and 2 were significantly different (P = .044).

DISCUSSION

We have found that the addition of an interference screw to a docking construct significantly improves the stiffness and the moment at 3 mm of gap formation, but does not significantly affect the ultimate moment of failure. Moreover, it appears that the interference screw improves the ability of the construct to resist static valgus loading.

When an interference screw was added to the reconstruction, the laxity, or ability to resist static valgus loading under our testing conditions, was no different than the intact state. When no screw was used, the reconstructed specimens demonstrated more significant laxity than the intact state. These findings may favor the use of screws; however, it is also possible that screws may overconstrain the elbow, especially at higher flexion angles as demonstrated in Figure 7. This may not be clinically significant because it is a time-0 study, and stress relaxation of the graft may counter the initial overconstraint.

McAdams et al[15] found increased valgus laxity in their docking specimens when compared with the intact specimens. However, this occurred at cycles 10 and 100 in their cyclic loading study, and no significant difference was found after 1 cycle. This is similar to our study, but strict comparisons are difficult to make given the differences in study design and sample size. Clinically, the residual laxity may be clinically insignificant due to a number of factors, including the effect of the dynamic stabilizers of the elbow.

Ahmad et al[1] reported an average stiffness for the reconstructed elbows to be 20.28 ± 12.5 N/mm. This was significantly less than their intact elbows (42.81 ± 11.6 N/mm). Our construct stiffness for group 2 (addition of interference screw) is comparable (14.7 ± 6.4 N/mm) and significantly different than our reconstructions without the screw. We did not measure the stiffness of our intact specimens out of concern of damaging structures or losing control of important variables that may have altered the subsequent experimental conditions. Although the Ahmad study provides a reference, it is difficult to strictly compare values due to the differences in experimental design (ie, metal vs

BioTenodesis screw and screw fixation on both ulna/humerus vs docking with humeral screw).

Our ultimate moment of failure for group 2, 14.4 ± 4.0 N·m, is consistent with others in the literature, but a strict comparison is impossible because of the variations in experimental technique. Ahmad et al[1] found in their study of interference screw reconstruction that the moment for intact elbows was 34.0 ± 6.9 N·m and, for reconstructed elbows, 30.6 ± 19.2 N·m at 70° of elbow flexion. Large et al[14] compared interference screw reconstruction with a Jobe reconstruction and intact ligament. The moment for their screw-reconstructed specimens at 70° of elbow flexion was 13.4 N·m, significantly less than the Jobe reconstruction (22.7 N·m). Paletta et al[18] tested quadrupled palmaris graft within a docking construct and found a moment to failure at 30° of flexion of 14.3 N·m for the docking and 18.8 N·m in the native ligament. In their study comparing intact specimens, specimens reconstructed with the Jobe bone-tunnel technique, and specimens with a suture anchor reconstruction, Hechtman et al[9] found an ultimate mode of failure to be 22.7 N·m for the intact, compared with 13.6 N·m for the bone anchor and 15.4 N·m for the bone tunnel. Like most studies in the literature, they tested at 70° of elbow flexion and were unable to match the failure strength of the intact ligament.[1,2,9,18] We were not able to detect a significant difference in moment to failure between the 2 groups. In addition, most specimens in both groups failed with the whipstitch pulling through the tendon. This diminishes the effect of the screw in ultimate failure loading of this construct. However, the moment at 3 mm of gap formation was significant. We chose to analyze this distance of gap formation because normal elbows can have 1 to 2 mm of medial opening. An incompetent UCL will demonstrate greater than 3 mm of opening between the coronoid and medial humerus.[7]

As mentioned above, multiple combinations (ie, metal vs bioabsorbable, cyclic vs noncyclic) and comparisons with interference screw fixation are being examined in UCL reconstruction. Theoretical advantages of interference screw fixation include a less technically demanding surgery, less dissection with less ulnar nerve morbidity, elimination of potential ulnar tunnel fracture, easier graft tensioning, greater reproducibility, and more isometric graft placement.[1] Each of these interference screw constructs employs a screw in both an ulnar and humeral tunnel. We believe that one of the strengths of our study is that it isolates humeral fixation. Furthermore, all specimens underwent identical testing in a pairwise fashion with strict control of experimental variables and, by inherent design in the study, factors such as muscle and joint compression were excluded. Lastly, the average age of our specimens was 49 years, which is older than the typical surgical population for this procedure, but improved when compared with other studies.[2,14,15,18] Although 2 ulnar fractures occurred, they were from the same donor and likely were a result of poor bone quality. A previously accepted experimental technique was used as model for our study design.[1]

This study has several limitations that should be considered when interpreting the results. Our study was similar to that of Ahmad et al[1] in that it simulated valgus torque with a static load. This does not replicate the dynamic nature of the acceleration forces necessary for overhead throwing. A higher number of specimens could have increased our power to show significant differences within our data. Also, this was a "time-0" study. The necessary strength and stiffness that are required of this graft construct for return to competitive throwing are not known. Ideally, the construct would be of equal or greater strength than the native ligament throughout the healing process. The effect of healing or scarring on the construct strength remains unknown. Certainly, the healing process could adversely affect the mechanical properties of the absorbable screws. Whether the healing process occurs at a sufficient rate to minimize these concerns is also unknown. Animal model studies will help us determine whether the bone tunnel or interference graft construct achieves greater strength and stiffness during and after healing of the graft.

Palmaris longus autograft, when present, is the most clinically used graft. Gracilis, toe extensors, and plantaris have been substituted in the absence of the palmaris longus.[3,6,13,19,20,22] The palmaris longus tendon was absent in 4 of our specimens, and we chose the FDS tendon as a substitute because of its availability. We noticed the average FDS tendon was thicker than the average palmaris tendon, and variability in the width of the palmaris grafts from specimen to specimen existed. Although efforts were made to match the size of the FDS tendon to the average palmaris, this factor was not strictly controlled. We observed that the interference screw was most beneficial when used with a smaller graft. Conversely, in the presence of a relatively large graft, the screw can overstuff the tunnel, potentially causing insertion difficulty or tunnel fracture.

Interestingly, although the angle used during failure testing was different, our ultimate moment to failure is similar to that of Paletta et al[18] (quadrupled palmaris graft) despite only using a doubled palmaris graft. We are currently conducting a study on the effect of tendon cross-sectional area on these constructs.

Although many modifications have been proposed, the rate of recovery has not changed considerably. One modification, the docking technique, has been increasingly used since the first results were reported in 2002. The authors advocated that the docking technique allowed tendon-to-bone healing within a tunnel (as opposed to side-to-side tendon healing), minimized humeral tunnels, and facilitated simpler graft tensioning.[20] In the first series, 33 of 36 patients (92%) returned to at least their previous level of competition for a 1-year period. When discussing the rehabilitation protocol, it was noted that "generally, players did not pitch in competitive situations until 9 months after the operation." A specific average interval of return to competition was not reported.[20]

A second series of 100 patients from the same primary author was published in 2006.[6] They reported excellent (90%) to good (7%) results with the Conway system.[5] Pitchers started a tossing program at 4 months, progressing to mound throwing at around 9 months. They were discouraged from competitive pitching until 1 year after surgery. Similar to their first series, the average interval of return to competition was not reported.

Although the docking technique alone is clinically successful at returning throwers to competition, it results in at least 1 full year lost from competition. We have demonstrated that the addition of an interference screw improves the initial stiffness and moment at 3 mm of gap formation. The addition of an interference screw to this construct may prove to be clinically significant, particularly to a smaller graft, and may allow patients to accelerate rehabilitation and ultimately return at a faster rate.

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